Document made available under the Patent Cooperation Treaty (PCT)

International application number: PCT/NZ05/000240

International filing date: 12 September 2005 (12.09.2005)

Document type: Certified copy of priority document

Document details: Country/Office: US
Number: 60/608,751
Filing date: 10 September 2004 (10.09.2004)

Date of receipt at the International Bureau: 21 November 2005 (21.11.2005)

Remark: Priority document submitted or transmitted to the International Bureau in compliance with Rule 17.1(a) or (b)

World Intellectual Property Organization (WIPO) - Geneva, Switzerland
Organisation Mondiale de la Propriété Intellectuelle (OMPI) - Genève, Suisse

PCT/NZ05 000240

PA 1368682

THE UNITED STATES OF AMERICA

TO ALL TO WHOM THESE PRESENTS SHALL COME:

UNITED STATES DEPARTMENT OF COMMERCE

United States Patent and Trademark Office

September 21, 2005

THIS IS TO CERTIFY THAT ANNEXED HERETO IS A TRUE COPY FROM THE RECORDS OF THE UNITED STATES PATENT AND TRADEMARK OFFICE OF THOSE PAPERS OF THE BELOW IDENTIFIED PATENT APPLICATION THAT MET THE REQUIREMENTS TO BE GRANTED A FILING DATE UNDER 35 USC 111.

APPLICATION NUMBER: *60/608,751*

FILING DATE: *September 10, 2004*

By Authority of the

Under Secretary of Commerce for Intellectual Property

and Director of the United States Patent and Trademark Office

M. K. Carter

M. K. CARTER

Certifying Officer

16085 U.S.PTO
091004

PTO/SB/16 (04-03)
Approved for use through 03/31/2003. OMB 0651-0032
U.S. Patent and Trademark Office; U.S. DEPARTMENT OF COMMERCE

PROVISIONAL APPLICATION FOR PATENT COVER SHEET

This is a request for filing a PROVISIONAL APPLICATION FOR PATENT under 37 CFR 1.53(c).

Express Mail Label No. EV374763290US

16800 U.S. PTO
60/608751
091004

INVENTOR(S)

Given Name (first and middle [if any])	Family Name or Surname	Residence (City and either State or Foreign Country)
RAY ANDREW	SIMPKIN	Auckland, New Zealand

Additional inventors are being named on the ______ *separately numbered sheets attached hereto*

TITLE OF THE INVENTION (500 characters max)

IMAGING SYSTEM

Direct all correspondence to: CORRESPONDENCE ADDRESS

[✓] Customer Number: 000110

OR

[ ] Firm or Individual Name	Vincent T. Pace
Address	Dann, Dorfman, Herrell and Skillman, P.C.
Address	1601 Market Street, Suite 2400
City	Philadelphia
State	PA
Zip	19103-2307
Country	US
Telephone	215-563-4100
Fax	215-563-4044

ENCLOSED APPLICATION PARTS *(check all that apply)*

[✓] Specification *Number of Pages* 26

[✓] Drawing(s) *Number of Sheets* 5

[ ] Application Data Sheet. See 37 CFR 1.76

[ ] CD(s), Number ______

[ ] Other (specify) ______

METHOD OF PAYMENT OF FILING FEES FOR THIS PROVISIONAL APPLICATION FOR PATENT

[ ] Applicant claims small entity status. See 37 CFR 1.27.

[✓] A check or money order is enclosed to cover the filing fees.

[✓] The Director is hereby authorized to charge filing fees or credit any overpayment to Deposit Account Number: 04-1406

[ ] Payment by credit card. Form PTO-2038 is attached.

FILING FEE Amount ($): $160.00

The invention was made by an agency of the United States Government or under a contract with an agency of the United States Government.

[✓] No.

[ ] Yes, the name of the U.S. Government agency and the Government contract number are: ______

Respectfully submitted,

SIGNATURE Vincent T. Pace

Date 09/10/2004

REGISTRATION NO. 31,049 *(if appropriate)*

TYPED or PRINTED NAME VINCENT T. PACE

Docket Number: 0074-501865

TELEPHONE 215-563-4100

USE ONLY FOR FILING A PROVISIONAL APPLICATION FOR PATENT

16085 U.S. PTO

PTO/SB/17 (10-03)
Approved for use through 07/31/2006. OMB 0651-0032
U.S. Patent and Trademark Office; U.S. DEPARTMENT OF COMMERCE
Under the Paperwork Reduction Act of 1995, no persons are required to respond to a collection of information unless it displays a valid OMB control number.

FEE TRANSMITTAL for FY 2004

Effective 10/01/2003. Patent fees are subject to annual revision.

☐ Applicant claims small entity status. See 37 CFR 1.27

TOTAL AMOUNT OF PAYMENT ($) 160.00

Complete if Known	
Application Number	not assigned yet
Filing Date	09/10/2004
First Named Inventor	RAY ANDREW SIMPKIN
Examiner Name	N/A
Art Unit	N/A
Attorney Docket No.	0074-501865

METHOD OF PAYMENT (check all that apply)

☑ Check ☐ Credit card ☐ Money Order ☐ Other ☐ None

☑ Deposit Account:

Deposit Account Number: 04-1406

Deposit Account Name: DannDorfmanHerrell&Skillm

The Director is authorized to: (check all that apply)

☐ Charge fee(s) indicated below ☑ Credit any overpayments

☑ Charge any additional fee(s) or any underpayment of fee(s)

☐ Charge fee(s) indicated below, except for the filing fee

to the above-identified deposit account.

FEE CALCULATION

1. BASIC FILING FEE

Large Entity Fee Code	Fee ($)	Small Entity Fee Code	Fee ($)	Fee Description	Fee Paid
1001	770	2001	385	Utility filing fee	
1002	340	2002	170	Design filing fee	
1003	530	2003	265	Plant filing fee	
1004	770	2004	385	Reissue filing fee	
1005	160	2005	80	Provisional filing fee	160.00

SUBTOTAL (1) ($) 160.00

2. EXTRA CLAIM FEES FOR UTILITY AND REISSUE

		Extra Claims		Fee from below		Fee Paid
Total Claims	-20** =		X		=	
Independent Claims	- 3** =		X		=	
Multiple Dependent					=	

Large Entity Fee Code	Fee ($)	Small Entity Fee Code	Fee ($)	Fee Description
1202	18	2202	9	Claims in excess of 20
1201	86	2201	43	Independent claims in excess of 3
1203	290	2203	145	Multiple dependent claim, if not paid
1204	86	2204	43	** Reissue independent claims over original patent
1205	18	2205	9	** Reissue claims in excess of 20 and over original patent

SUBTOTAL (2) ($)

**or number previously paid, if greater; For Reissues, see above

FEE CALCULATION (continued)

3. ADDITIONAL FEES

Large Entity Fee Code	Fee ($)	Small Entity Fee Code	Fee ($)	Fee Description	Fee Paid
1051	130	2051	65	Surcharge - late filing fee or oath	
1052	50	2052	25	Surcharge - late provisional filing fee or cover sheet	
1053	130	1053	130	Non-English specification	
1812	2,520	1812	2,520	For filing a request for *ex parte* reexamination	
1804	920*	1804	920*	Requesting publication of SIR prior to Examiner action	
1805	1,840*	1805	1,840*	Requesting publication of SIR after Examiner action	
1251	110	2251	55	Extension for reply within first month	
1252	420	2252	210	Extension for reply within second month	
1253	950	2253	475	Extension for reply within third month	
1254	1,480	2254	740	Extension for reply within fourth month	
1255	2,010	2255	1,005	Extension for reply within fifth month	
1401	330	2401	165	Notice of Appeal	
1402	330	2402	165	Filing a brief in support of an appeal	
1403	290	2403	145	Request for oral hearing	
1451	1,510	1451	1,510	Petition to institute a public use proceeding	
1452	110	2452	55	Petition to revive - unavoidable	
1453	1,330	2453	665	Petition to revive - unintentional	
1501	1,330	2501	665	Utility issue fee (or reissue)	
1502	480	2502	240	Design issue fee	
1503	640	2503	320	Plant issue fee	
1460	130	1460	130	Petitions to the Commissioner	
1807	50	1807	50	Processing fee under 37 CFR 1.17(q)	
1806	180	1806	180	Submission of Information Disclosure Stmt	
8021	40	8021	40	Recording each patent assignment per property (times number of properties)	
1809	770	2809	385	Filing a submission after final rejection (37 CFR 1.129(a))	
1810	770	2810	385	For each additional invention to be examined (37 CFR 1.129(b))	
1801	770	2801	385	Request for Continued Examination (RCE)	
1802	900	1802	900	Request for expedited examination of a design application	

Other fee (specify) ____

*Reduced by Basic Filing Fee Paid

SUBTOTAL (3) ($)

SUBMITTED BY			(Complete (*if applicable*))		
Name (Print/Type)	Vincent T. Pace	Registration No. (Attorney/Agent)	31,049	Telephone	215-563-4100
Signature	Vincent T. Pace			Date	09/10/2004

WARNING: Information on this form may become public. Credit card information should not be included on this form. Provide credit card information and authorization on PTO-2038.

This collection of information is required by 37 CFR 1.17 and 1.27. The information is required to obtain or retain a benefit by the public which is to file (and by the USPTO to process) an application. Confidentiality is governed by 35 U.S.C. 122 and 37 CFR 1.14. This collection is estimated to take 12 minutes to complete, including gathering, preparing, and submitting the completed application form to the USPTO. Time will vary depending upon the individual case. Any comments on the amount of time you require to complete this form and/or suggestions for reducing this burden, should be sent to the Chief Information Officer, U.S. Patent and Trademark Office, U.S. Department of Commerce, P.O. Box 1450, Alexandria, VA 22313-1450. DO NOT SEND FEES OR COMPLETED FORMS TO THIS ADDRESS. **SEND TO: Commissioner for Patents, P.O. Box 1450, Alexandria, VA 22313-1450.**

If you need assistance in completing the form, call 1-800-PTO-9199 and select option 2.

IMAGING SYSTEM

FIELD OF THE INVENTION

The present invention relates to an imaging system for body parts which utilises non-ionizing electromagnetic radiation, for example microwaves. In particular, although not exclusively, the imaging system is suitable for breast cancer screening.

BACKGROUND TO THE INVENTION

Breast cancer is the most common cancer to affect women. The detection of malignancies at an early stage is deemed to offer the best prognosis for patients and this has lead to the establishment of screening programmes aimed at early detection.

X-ray mammography is one commonly used breast cancer screening method due to its simplicity, high-resolution images and cost effective implementation. However, x-ray mammography has a number of associated limitations and drawbacks. X-rays are an example of ionizing electromagnetic radiation which can damage tissue and in some cases initiate malignant tumours. X-ray mammography requires the patient's breasts to be compressed between two plates which is uncomfortable for many women and makes it difficult to determine the true three-dimensional (3D) location of any suspicious features. Furthermore, women with silicone breast implants are also at risk from implant rupture due to the compression process. X-ray images are two-dimensional (2D) and a number of images from different views must typically be taken to provide some indication of the 3D location of suspicious features. X-ray detection of suspicious features relies on differences in density within the breast tissue under test and the density contrast between healthy and malignant breast tissue is small, typically only about 2%, which can make detection of tumours difficult. For post-menopausal women, x-ray mammography fails to detect up to 15% of cancers. For younger women, whose breast density is usually higher, up to 40% of cancers can be missed by x-ray mammography. Generally, the smallest tumour detectable with x-ray mammography is

about 4mm in diameter. A tumour this size is reckoned to have been in the body for about 6 years, that is, not particularly early in the tumour's development.

All of the above have provided significant incentive for researchers to develop alternative methods for breast cancer detection which alleviate some of the difficulties associated with x-ray mammography. Radar imaging, which utilises electromagnetic waves in the microwave region, has been identified as having potential for improved detection of breast cancer due to the large difference in complex permittivity between healthy and malignant breast tissue. US patent numbers 4,641,659, 5,807,257, 5,829,437, 6,448,788, and 6,504,288 disclose various radar breast imaging systems.

It is an object of the present invention to provide an improved imaging system for body parts, or to at least provide the public with a useful choice.

SUMMARY OF THE INVENTION

In one aspect, the present invention broadly consists in a method for generating a three-dimensional image of a body part including the steps of:

scanning to obtain surface profile information for the body part;

transmitting non-ionizing radiation having a frequency or frequencies above approximately 10GHz into the body part and then receiving non-ionizing radiation reflected back from the body part at a multiplicity of predetermined locations defining a synthetic aperture relative to the body part, the radiation being transmitted and received at each of the predetermined locations by moving one or more antenna elements within the synthetic aperture and sequentially operating the or each antenna element;

obtaining radiation information at each of the predetermined locations from the reflected radiation received; and

processing the surface profile information and radiation information to generate a three-dimensional image of the body part, the three-dimensional image having a plurality of image points.

Preferably, the method further includes transmitting and receiving radiation at more than one frequency at each of the predetermined locations to obtain additional radiation information. More preferably, the method includes transmitting and receiving microwave radiation at a plurality of discrete frequencies between 10GHz and 18GHz.

Preferably, processing the surface profile information and radiation information includes synthetically focusing the radiation information to each image point of the three-dimensional image of the body part. More preferably, synthetically focusing the radiation information to each image point includes coherently adding radiation information from each of the predetermined locations.

Preferably, the method includes transmitting non-ionizing radiation through air toward the body part and receiving non-ionizing radiation reflected back through air from the body part at each of the predetermined locations.

Preferably, the method includes moving an array of antenna elements within the synthetic aperture and sequentially operating each antenna element to transmit and receive non-ionizing radiation at each of the predetermined locations. Alternatively, the method includes moving a single antenna element within the synthetic aperture and operating it to transmit and receive non-ionizing radiation at each of the predetermined locations.

Preferably, obtaining radiation information at each of the predetermined locations includes measuring the reflection coefficient of the reflected radiation received. More preferably, this includes measuring the amplitude and phase of the reflection coefficient.

Preferably, scanning to obtain surface profile information for the body part includes operating a three-dimensional laser profiler. More preferably, the method includes obtaining the surface profile information and radiation information simultaneously.

Preferably, the method generates a three-dimensional radar image of the body part, which may be, for example, a human breast.

In another aspect, the present invention broadly consists in an imaging system for generating a three-dimensional image of a body part including:

a three-dimensional profiler arranged to scan the body part and obtain surface profile information;

a radar device arranged to transmit non-ionizing radiation having a frequency or frequencies above approximately 10GHz into the body part and then receive non-ionizing radiation reflected back from the body part at a multiplicity of predetermined locations defining a synthetic aperture relative to the body part to thereby obtain radiation information at each of the predetermined locations, the radar device having one or more antenna elements which are moveable within the synthetic aperture and operable to transmit and receive the non-ionizing radiation; and

a control system arranged to operate the three-dimensional profiler and radar device, and which processes the surface profile information and radiation information to generate a three-dimensional image of the body part, the three-dimensional image having a plurality of image points.

Preferably, the radar device includes a radiation source and radiation receiver associated with the antenna element(s), the radiation source being operable to generate non-ionizing radiation at more than one frequency. More preferably, the radar device is operable to obtain radiation information at more than one frequency at each of the predetermined locations. In the preferred form system, the radiation source is a microwave source which generates microwaves at a plurality of discrete frequencies between 10GHz and 18GHz.

Preferably, the control system includes an imaging system arranged to process the surface profile information and radiation information to generate the three-dimensional image of the body part. More preferably, the imaging system is arranged to synthetically focus the radiation information to each image point of the three-dimensional image of the body part. Even more preferably, the imaging system is arranged to synthetically focus the radiation information by coherent addition of the radiation information from each of the predetermined locations within the synthetic aperture.

Preferably, the antenna element(s) of the radar device are displaced from the body part and are arranged to transmit non-ionizing radiation through air toward the body part and receive non-ionizing radiation reflected back through air from the body part.

In the preferred form, the radar device includes an array of antenna elements moveable by an operable scanning mechanism, each antenna element of the array being selectively connectable to the radiation source and receiver via operation of a switching network. Preferably, the control system is arranged to operate the scanning mechanism and switching network to move the array within the synthetic aperture and sequentially operate each antenna element to obtain radiation information at each of the predetermined locations within the synthetic aperture.

In an alternative form, the radar device includes a single antenna element moveable by an operable scanning mechanism, the antenna element being connected to the radiation source and receiver. Preferably, the control system is arranged to operate the scanning mechanism to move the antenna element to each predetermined location within the synthetic aperture for operation to obtain the radiation information.

Preferably, the radiation receiver is arranged to obtain radiation information by measuring the reflection coefficient of the reflected radiation received by the antenna element(s). More preferably, the radiation receiver is arranged to measure the amplitude and phase of the reflection coefficient.

Preferably, the three-dimensional profiler is laser-based. Alternatively, the three-dimensional profiler is arranged to obtain surface profile information by utilising ultrasound or broadband microwave signals.

Preferably, the control system is arranged to operate the three-dimensional profiler and radar device to obtain the surface profile information and radiation information simultaneously.

Preferably, the control system is arranged to generate a three-dimensional radar image of the body part, which may be, for example, a human breast.

The invention may also be said broadly to consist in the parts, elements and features referred to or indicated in the specification of the application, individually or collectively, and any or all combinations of any two or more said parts, elements or features, and where specific integers are mentioned herein which have known equivalents in the art to which this invention relates, such known equivalents are deemed to be incorporated herein as if individually set forth.

The invention consists in the foregoing and also envisages constructions of which the following gives examples only.

BRIEF DESCRIPTION OF THE DRAWINGS

Preferred forms of the invention will be described by way of example only and with reference to the drawings, in which:

Figure 1 is a rendered perspective view of a preferred form breast imaging system having a sensor head attached to a robot scanner;

Figure 2 is a rendered perspective view of the sensor head of Figure 1;

Figure 3 is a block diagram of the preferred form breast imaging system;

Figure 4 is a block diagram of the radar device of the breast imaging system; and

Figure 5 is a schematic diagram showing the geometry relevant to the synthetic focusing algorithm.

DETAILED DESCRIPTION OF PREFERRED EMBODIMENTS

The preferred form imaging system of the invention is a breast cancer screening tool and is arranged to scan a patient's breasts with microwave radiation in order to generate 3D radar images of each breast which can be examined for suspicious features such as malignant tumours. There is a large difference in complex permittivity between healthy and malignant breast tissue and this leads to greater scattered field amplitudes from malignant tumours embedded in healthy tissue which show up readily in a microwave image of scattered field intensity. For example, the real part of complex permittivity (the dielectric constant) for a malignant tumour is of the order of 50 at a frequency of 10GHz whereas healthy tissue has a value of about 9. Hence, radar images are suited to breast tumour detection since the high permittivity contrast between malignant and healthy tissue translates to high-contrast images.

The imaging system generates 3D radar images based on the intensity of the scattered field as a function of position from measurements of scattered fields external to the breast. The imaging system utilises a focusing algorithm to provide coherent addition of scattered fields at a given image point within the 3D radar image, thereby giving a measure of the scattered field intensity at a point in the breast being scanned.

Referring to Figure 1, the preferred form imaging system 100 includes a sensor head 101 which is translated relative to a patient 102 by a robot 103. The imaging system is arranged to scan each of the patient's exposed breasts individually and generate respective 3D radar images. In particular, the imaging system scans the patient's breasts to simultaneously obtain radiation information and surface profile information which are processed by an image generation algorithm to generate the 3D radar images. The preferred form sensor head 101 does not make contact with the patient 102 and there is no coupling medium, other than air, between the patient and sensor head during scanning. In an alternative form the sensor head 101 could be moved by means other than robot 103, or alternatively the patient could be moved relative to the sensor head.

Referring to Figure 2, the sensor head is mounted to the robot scanning mechanism in the preferred form by a mounting flange 200. The sensor head includes a 3D profiler 201 which is arranged to obtain geometric surface profile information of the breast during scanning. In the preferred form, the 3D profiler is a laser profiler device which uses a scanning laser stripe and charge-couple device (CCD) sensor to provide range information by triangulation. The laser output power from the 3D profiler is deemed eye-safe. It will be appreciated that other types of 3D profiling devices could be utilised to obtain surface profile information of the breasts. For example, alternative forms of 3D profilers may utilise ultrasound or broadband microwave signals to obtain the surface profile information.

The sensor head also carries a radar device which is arranged to transmit non-ionizing radiation toward the breast and then receive radiation reflected back from the breast at a plurality of predetermined locations relative to the breast. The radar device includes a radiation source 202 and receiver 203 which are connected to an array 204 of antenna elements or waveguides via a switching network 205. In the preferred form, the radiation source 202 is a Yttrium Iron Garnate (YIG) oscillator which generates microwaves over a broad range of frequencies and the radiation receiver 203 is a six-port reflectometer. The radar device is operated and controlled by an on-board computer system 206 and also has a calibration device 207 and an associated servo-motor 208.

The preferred form radar device is arranged to obtain radiation information from a multiplicity of predetermined locations which define a synthetic aperture relative to the breast. The radar device sweeps out the synthetic aperture by translating the antenna array 204 within the synthetic aperture and sequentially operating each of the individual antenna elements. For example, the preferred form radar device has a linear array of thirty two antenna elements arranged in two rows of sixteen antenna elements. During scanning, the antenna array is, for example, translated mechanically by the robot scanning mechanism to thirty two equally spaced locations in an orthogonal direction relative to the antenna array. At each of the thirty two locations, the thirty two individual antenna elements are sequentially connected to the radiation source and receiver by the switching network so that radiation information can be obtained at each

of the 1024 locations of the synthetic aperture. In the preferred form, the antenna array has monostatic antenna elements, i.e. the antenna elements both transmit and receive microwave signals, but separate transmittal and receive elements could be used in an alternative arrangement.

The size of the synthetic aperture should preferably be no less than twice that of the body part to be imaged, so that the body part is illuminated sufficiently well by electromagnetic radiation from each antenna element. For imaging a breast, a value of 15cm has been assumed as a typical linear dimension. Therefore, the minimum synthetic aperture size, D, is preferably twice this value, namely 30cm along each transverse axis. It will be appreciated that imaging system can alternatively operate with a smaller synthetic aperture to body part ratio depending on the system requirements.

The required antenna element spacing in the antenna array is determined from the requirement to satisfy the Nyquist sampling criterion at the highest frequency of operation (shortest wavelength) so that grating lobes are avoided in the resulting image. This criterion requires that the element spacing be no greater than one half of a wavelength at the highest frequency of operation. For example, an upper frequency limit of 18GHz gives the largest allowed element spacing as 8.3mm. This element spacing in turn dictates the number of predetermined antenna locations in the synthetic aperture when combined with the minimum synthetic aperture size.

The radiation information at each location within the synthetic aperture is obtained by illuminating the breast with microwave radiation from a transmitting antenna and then measuring the amplitude and phase of the reflected wave (scattered field) from the breast. The microwave radiation may be at a single frequency or small number of frequencies but in the preferred form radiation information is obtained at each location by repeating the measurement over a broad range of frequencies, one frequency at a time. In the preferred form radar device, a six-port reflectometer is incorporated into the microwave signal path. The six-port reflectometer is arranged to produce four voltages from diode detectors connected to its output ports from which it is possible to determine

the amplitude and phase of the reflected signals relative to the incident (transmitted) signal.

It will be appreciated that there are other alternative antenna arrangements which could be utilised to obtain the radiation information at each of the locations within the synthetic aperture. For example, the radar device may be equipped with only a single antenna element which is translated mechanically to all locations with the synthetic aperture, although such an arrangement would be slow in terms of data acquisition speed. Alternatively, a real aperture could be provided in which there is an antenna element at each of the predetermined locations over the breast. With a fixed, real aperture the radiation information would be obtained by sequentially operating each antenna element one at a time. While a real aperture arrangement would be fast from a data acquisition viewpoint, it would also be more costly. The preferred form radar device is a combination of these two alternatives and is a compromise between data acquisition time and cost.

Referring to Figure 3, the sensor head 300 is mounted to a robot scanning mechanism 301 and carries both the 3D profiler 302 and radar device 303. The robot scanning mechanism 301 is arranged to move the sensor head 300 relative to a patient's breast while the 3D profiler 302 and radar device 303 obtain surface profile information and radiation information respectfully as described above. A control system 304 is provided which controls the robot scanning mechanism 301, 3D profiler 302 and radar device 303 during the breast scan. Further, the control system 304 is arranged to process the surface profile and radiation information with an imaging system to generate the 3D radar image of the breast.

Referring to Figure 4, the configuration and operation of the radar device 303 will be explained in more detail. The radar device 303 communicates with the control system 304 via an on-board computer system 400. The radar device has a YIG oscillator 401 which is operated in a swept frequency mode via its driver circuit 402 to generate microwave radiation at the desired discrete frequencies. The driver circuit 402 is in turn controlled by a sequence of binary signals from the on-board computer system 400.

An important feature of the radar device is that the microwave power level emitted by each antenna element in the antenna array 403 is low and is of a non-ionizing nature. For example, the microwave power output from the YIG oscillator 401 may vary from 30mW-50mW depending on the frequency. However, the power level made available to each radiating element in the antenna array 403 may be in the order of 0.1mW due to attenuation in the six-port reflectometer 404 and switching network 405. The sensor head 303 is also displaced, for example approximately 30cm, away from the patient's body which further reduces radiation exposure to the patient. Therefore, from a radiological stand point, the radar device is inherently safe.

Also, this stand-off distance is not critical but should preferably be greater than five wavelengths at the lowest frequency of operation in order that the illuminating wavefront from each antenna element has a spherical phase front with local plane-wave characteristics. That is, the breast is far removed from the reactive near-field region of the antenna and is illuminated by a wavefront having predictable phase and amplitude characteristics. A stand-off distance of ten wavelengths at the lowest frequency of operation is most preferable for reducing the effects of multiple reflections between breast and antenna which can contaminate the measured data and subsequent radar images. The stand-off distance is a compromise between being large enough to satisfy the above criteria and small enough that the transmitted and received signal levels are not too low due to the space-attenuation factor (that is the $1/R^4$ dependence on the received power level, R being the object-antenna separation). This effect is compensated for in the preferred form by using a large number of elements in the synthetic aperture to enhance the received power levels when applying synthetic focussing. In addition, during the synthetic focussing process (which will be subsequently described), the size of the focal spot is also degraded (i.e. becomes larger) as the object-antenna separation is increased. To this end, it is desirable to maintain a focal ratio of the order of unity in determining the appropriate stand-off distance.

A non-contact sensor head 300 enables the reflected signals from the breast to be accurately measured and allows calibration of the antenna system of the radar device in

isolation. As mentioned, the stand-off distance between the breast and the plane of the synthetic aperture should preferably be at least 10 wavelengths at the lowest frequency of operation in order to reduce the effects of multiple reflections between antenna and breast to a negligible level. This allows the effects of the antenna system to be subtracted from the measured radiation information with the breast in place to give just the reflectivity of the breast in isolation. A typical stand-off distance used for the breast imaging device is therefore 30cm at a minimum operating frequency of 10GHz.

The radiation information to be measured by the radar device is the reflection coefficient of the reflected microwave signals at each location within the synthetic aperture and at each frequency of interest. In particular, the phase and amplitude of the reflection coefficient is measured. The six-port reflectometer 404 within the microwave signal path produces four voltages from diode detectors connected to its output ports from which the amplitude and phase of the reflected signals relative to the incident (transmitted) signal is determined.

The six-port reflectometer 404 essentially combines the reflected microwave signal from the breast under test with a portion of the incident wave. This is done using four different relative phase differences introduced by the six-port reflectometer 404 between incident and reflected waves. The four combinations of microwave signals are then sent to four square-law detector diodes which generate four output voltages. One of the four output voltages is used as a reference such that three voltage ratios are derived for each measurement. These three ratios are converted into the real and imaginary parts of the reflection coefficient. The measured reflection coefficient information is then converted into digital data by an analogue-to-digital converter which in turn sends the digital data to the on-board computer system 400.

The radar device employs a near-field imaging method in that the distance between the antenna elements and the patient's breast has a focal ratio typically in the order of unity. Therefore, the transmitted wavefronts illuminating the breast are highly curved. Further, the imaging system utilises an image generation algorithm which images objects embedded in the breast interior. In particular, the image generation algorithm takes into

account the refraction at the various dielectric interfaces in order to focus effectively within the breast.

In the preferred form the radar device 303 includes a calibration device 407 and associated servo-motor 408 which are arranged to calibrate the six-port reflectometer 404 and antenna system. Calibration of the six-port reflectometer 404 will be described first. In order to accurately determine the complex reflection coefficient from the voltage outputs of the six-port reflectometer 404, it is necessary to calibrate the reflectometer to account for any imperfections and idiosyncrasies in the componentry. A number of 'calibration standards' are connected to the measurement port of the reflectometer and output voltages acquired as per a normal measurement. The calibration standards have known reflection coefficients for all frequencies of interest. For example, for the preferred form radar device, nine standards are used, all of them different lengths of short-circuited rectangular waveguide.

It is possible to calibrate a six-port reflectometer using only five standards. However, a total of nine are made available in the preferred form imaging system due to the broad range of frequencies used. The key to an accurate calibration procedure for a six-port reflectometer is the selection of five standards with widely spaced reflection coefficient phase angles (the magnitude of the reflection coefficient is unity for all short-circuit standards). Having nine standards available allows one to select the five best phase angles for use at a given frequency thereby maintaining accurate calibration across the whole frequency band.

The waveguide standards are built into the rotary calibration device 407 mounted on the sensor head which is able to connect each standard to the reflectometer measurement port one at a time by means of a servo-motor 408 to the appropriate position.

For each of the nine calibration standards, the four six-port reflectometer output voltages are measured for each frequency in the band and stored. These are converted into real and imaginary parts of reflection coefficient and a set of calibration coefficients generated using a standard algorithm (not described here). The calibration

coefficients characterise the six-port reflectometer 404 and enable the reflection coefficient of a breast under test to be accurately determined from the four diode detector output voltages taking into account the imperfections in the reflectometer 404 itself.

The calibration of the antenna system will now be described. In order to extract the amplitude and phase of the reflection coefficients attributable purely to the patient's breast under test it is necessary to remove the contribution from the antenna system. This is done by performing a series of reflection coefficient measurements on the antenna system with no patient present. In particular, two measurements are carried out on the antenna system as outlined below.

First measurement: With no patient present, the antenna system is positioned by the robot scanning mechanism so as to radiate into free-space with no reflective objects within close range. Each antenna element in the linear array is switched on in turn and the reflection coefficient determined for all frequencies via the output voltages from the six-port reflectometer. This represents the complex reflection coefficient of the antenna system and its associated switching network components and is referred to as the 'empty room' case. The most significant contribution to the reflection coefficient in this case will be from the antenna apertures.

Second measurement: The procedure outlined above is repeated with a metallic plate placed in close contact with the apertures of each antenna element in the linear array. This is referred to as the 'short circuit' case. The robot scanner moves the antenna array to a position where a metal plate is automatically in close contact with the aperture plane. The most significant contribution to the reflection coefficient in this case will be from the short circuit plate.

The short-circuit and empty-room data is used to extract the reflection coefficient of the breast alone from the overall measured reflection coefficient using the antenna array. This is an example of 'de-embedding' applied to the measured reflection coefficient

data to determine the reflection coefficient of the object in isolation. The de-embedding equation used is as follows:

$$R_{breast} = \frac{R_{meas} - R_{empty}}{R_{empty} - R_{short}} \qquad \ldots(1)$$

where

R_{breast} = Reflection coefficient of breast in isolation.
R_{meas} = Reflection coefficient of antenna and breast combined.
R_{empty} = Reflection coefficient of isolated antenna radiating into free-space.
R_{short} = Reflection coefficient of antenna with short-circuit applied across apertures.

The YIG oscillator 401 of the radar device generates continuous wave (CW) electromagnetic radiation covering a broad frequency bandwidth, i.e. the preferred form imaging system operates in broadband. In the preferred form, the operating frequency band is from 10GHz to 18GHz and radiation information is acquired at a number of frequencies throughout the band at each location within the synthetic aperture. The broadband frequency domain operation is utilised in order to provide a small focal spot size and hence good image resolution in the down-range direction. In the preferred form radar device, 161 discrete frequencies are used corresponding to a frequency interval of 50MHz between 10GHz and 18GHz. The frequency interval is chosen to be small enough such that aliasing in the down-range direction is avoided in the final 3D radar images for the locations of interest in the image space.

In order to obtain good focusing properties that approach the theoretical diffraction limit of half a wavelength for the size of the focal spot in the transverse plane, the synthetic aperture size needs to be large compared to the wavelength, λ. Therefore, the requirement that D = 10λ at the lowest frequency (longest wavelength) follows. If D = 30cm as mentioned previously, then λ = 3cm. Therefore, the minimum frequency of operation for the imaging system is preferably 10GHz.

The broader the frequency bandwidth, the better the down-range resolution, so as broad a bandwidth as possible is desirable. Following on from above, 20GHz is preferably the upper frequency of operation. However, the vast majority of components will only work over a limited band, typically an octave at best. Therefore 18GHz is more practical from the performance of available components, giving a bandwidth of 8GHz.

The frequency interval between steps as the device is swept across the full frequency band is also determined by the need to satisfy the Nyquist sampling criterion. A small enough frequency interval needs to be used so as to avoid grating lobes in the time domain response resulting from an integration over the frequency domain data. This is in turn related to the round-trip time delay from source to receiver via the object under test. The frequency interval is chosen so that alias bands in the time domain response do not lie within the time interval for signals to make a round trip. This time delay can also be represented as an equivalent distance (there and back) in free-space referred to as the Alias-Free Range (AFR). A frequency interval of 50MHz is used in the preferred form breast imaging system giving 161 frequencies between 10GHz and 18GHz.

Denoting the frequency interval by δf, the corresponding separation of alias bands in the time domain, δt, is given by the following equation:

$$\delta t = \frac{1}{\delta f} \qquad \text{...(2)}$$

Equation (2) can be used to calculate an equivalent 'round-trip' distance in free-space, (AFR), by multiplying δt by 2c where c is the speed of light in free-space to give equation:

$$AFR = 2c\delta t = \frac{2c}{\delta f} \qquad \text{...(3)}$$

The microwave path length between source and image point and back should be less than the AFR in order to avoid contamination of the radar images from alias responses

due to the sampling interval used in the frequency domain. Using δf = 50MHz in equation (3) gives AFR = 11.99 meters in free space which is deemed to be sufficiently large for the proposed imaging system to avoid alias responses. A larger number of frequencies (and therefore a smaller frequency interval) could be used but this has to be offset against the total data acquisition time which must be kept small so as not to inconvenience the patient. For example, the patient should ideally be able to hold there breath for the duration of the scan.

The system described thus far gathers reflection coefficient data (radiation information) from the breast over a range of microwave frequencies. Images of scattered field intensity (3D radar images) are then generated by applying a synthetic focussing algorithm.

Figure 5 shows the geometry of the antenna and breast configuration in a 3D Cartesian coordinate system. Referring to Figure 5 the vector $\mathbf{R}_1$ extends from the point in the antenna measurement plane to a point on the outer surface of the breast denoted by $P_s(x_s,y_s,z_s)$. The vector $\mathbf{R}_2$ extends from this point on the outer skin surface to a point on the interior skin surface. The vector $\mathbf{R}_3$ extends from this interior skin surface point to the image point $P'(x',y',z')$, the point at which microwave energy is to be focussed. This image point can be chosen arbitrarily. However, the path mapped out by the vectors $\mathbf{R}_1$, $\mathbf{R}_2$ and $\mathbf{R}_3$ between antenna point and image point is not defined in an arbitrary fashion. Fermat's Principle is invoked so that the optical path is the minimum one possible. The minimum optical path, R_{min}, is defined as follows for the geometry of Figure 5:

$$R_{\min} = \text{Minimum Value of } \left\{ |\mathbf{R}_1| + \sqrt{\varepsilon_{skin}}\,|\mathbf{R}_2| + \sqrt{\varepsilon_{tissue}}\,|\mathbf{R}_3| \right\} \qquad \ldots(4)$$

where

ε_{skin} = Dielectric constant of skin.

ε_{tissue} = Dielectric constant of breast tissue.

There is one minimum path R_{min} for each image point and antenna point. So, for a given point in the image, there is a set of N R_{min} values where N is the number of antenna points used in the synthetic aperture.

The scattered electric field vector measured by the antenna at the point P(x,y,z) at a frequency denoted by the free-space propagation constant, k, is defined as $\boldsymbol{E}_{scat}(x,y,z,k)$. The free-space propagation constant, k, is given by $2\pi/\lambda$ where λ is the free-space wavelength. A planar synthetic aperture is used here so that z = constant on the measurement plane.

The radar image at a given point P' is now formed by applying a phase shift equal to $2kR_{min}$ to the measured reflection coefficient data for each point in the synthetic aperture and then summing over all antenna locations. Summing over the frequency domain is also carried out. If the dielectric properties of the skin and breast tissue are assumed to vary negligibly with frequency (which is a good approximation), then the minimum paths between each image point and all antenna points will not depend on frequency. Therefore, once the minimum paths have been computed for a given combination of image point and antenna points, they can be used for all frequencies in the summation over the frequency domain.

Mathematically, the above process can be represented by the following three-fold integral for generating the image, I, at P'(x',y',z'):

$$I(x',y',z') = \iint_S \int_{k_1}^{k_2} E_{scat}(x,y,z,k)e^{2jkR_{min}}\,dSdk \qquad \ldots(5)$$

where

S = Synthetic aperture area.
k_1 = Free-space propagation constant at lowest frequency.
k_2 = Free-space propagation constant at highest frequency.

In equation (5) the factor of 2 in the phase shift term is present due to the need to account for the two-way path 'there and back' between antenna and image point. This phase shift term equalises the phase of the received signals from a given image point at all antenna locations so that when the summation over the synthetic aperture takes place, all quantities add up in phase to produce a much enhanced field at the image point location. The measured fields are therefore focussed at the image point. This is an example of synthetic focussing applied to an antenna array.

The use of the minimum optical path R_{min} to calculate the appropriate phase shift is consistent with the Method of Stationary Phase often used to evaluate integrals of the type given in equation (5). This type of integral is characterised by a phase function in the integrand - often expressed as a complex exponential like that in (5) - which is a function of the integration variables. For values of the phase function which are varying rapidly with position, the oscillatory nature of the integrand in these regions results in a negligible contribution to the integral since positive and negative going portions of the oscillatory phase function tend to cancel each other out. The only significant contribution to the value of the integral comes from the region where the phase function is varying slowly such as in the vicinity of a stationary point in the phase function. This region corresponds to the minimum path R_{min} and this is why it is used in the phase function $\exp(2jkR_{min})$ of the integrand in (5).

The vector nature of the electric field in (5) has been ignored since the dominant scattered field component will be co-polarised with the dominant polarisation present in the aperture of the antenna. That is, de-polarisation effects are ignored in the focussing algorithm – these will not be significant for a monostatic reflection coefficient measurement system.

Equation (5) appears simple in form but complexity lies in the need to determine the values of R_{min} for each combination of image point and antenna point. The determination of R_{min} can be performed as a separate computational exercise and need

only be computed once for a given antenna and breast geometry. In order to determine R_{min}, it is necessary to have knowledge of the following:

- the geometric profile of the breast's outer surface relative to some known origin.
- an estimate of the dielectric constants of the skin and interior breast tissue.
- an estimate of the skin thickness.

In the preferred form system, the geometric profile of the breast's outer surface is measured by the 3D laser profiler 201 co-mounted onto the radar sensor as described previously. Knowledge of skin thickness and dielectric constant of the skin and breast tissue to a high degree of accuracy is not necessary. An accepted value for the dielectric constant of skin at frequencies in the range 10GHz to 18GHz is 40 and that of the interior breast tissue is 9. The skin thickness may be nominally taken as 2mm. Values within 10% of the true values for dielectric constant will give rise to 5% errors in the optical path calculation due to the square-root dependence on the dielectric constants (see equation (4)).

For imaging purposes, the breast interior is assumed to be a homogeneous medium with a (mean) dielectric constant of ε_{tissue}. While the breast interior will not be homogenous in practice, deviations from this mean dielectric constant will not be large for normal breast tissue. Large deviations from this 'background' dielectric constant - such as encountered with malignant tumours – will show up readily in the radar image whereas the smaller deviations in dielectric properties normally encountered with healthy breast tissue will scatter weakly and not show up as significant features in the radar image. Typically the imaging system of the invention will operate as a breast screening tool aimed at detecting the presence of suspicious objects within the breast rather than as a diagnostic tool. The above assumption of homogeneity for the breast interior is deemed sufficient for screening purposes.

However, it is possible to glean additional information about the skin thickness and dielectric properties of the skin and breast interior from the focussed radar image data. The reflection coefficient due to the skin can be measured by focussing onto a region in

the skin's vicinity and comparing the amplitude and phase of the measured result with that of a theoretical transmission line model. This is carried out at a single frequency. Data for multiple frequencies can then be generated in a similar manner and further used to determine the thickness of the skin, its dielectric constant and that of the breast interior (assumed homogeneous). Focussing at the skin surface helps to eliminate contributions from the chest wall giving values of reflection coefficient for the skin region only.

The minimum path R_{min} is a function of the breast geometry as well as the antenna geometry and will therefore be unique to a particular patient. Values of R_{min} are calculated by fixing the antenna and image point locations and varying the position of the point P_s on the skin's outer surface until the minimum value of the optical path is found. The two variables of interest here are x_s and y_s ,the x and y coordinates on the outer surface of the skin. The value of z_s is governed by the outer surface profile data (as measured by the laser system) and is a function of x_s and y_s.

For a given point on the skin's outer surface, the point on the inner surface of the skin (where it meets the interior breast tissue) is automatically defined by Snell's Law of Refraction and so the vectors $\mathbf{R}_1$,$\mathbf{R}_2$ and $\mathbf{R}_3$ are all fully defined for given values of antenna and image points along with values of x_s and y_s. Snell's Law of Refraction is wholly consistent with Fermat's Principle for a minimum optical path. Thus, the only variables in the search routine for the minimum path are x_s and y_s.

Once found, the values of minimum path R_{min} are stored in a five-dimensional array. Two indices are used to define the antenna location in the synthetic aperture and a further three to define the image point in 3D space. Image generation then proceeds by the numerical evaluation of the integral in equation (5). The image itself is usually displayed as the magnitude of the image function I(x',y',z').

Use of commercially available 3D visualisation software is the most effective means of displaying the 3D radar image data. Iso-surfaces and volume rendering visualisations are particularly appropriate for detecting suspicious features within the breast.

The synthetic aperture method and apparatus described above consisting of an array of smaller antenna elements which behave collectively like an antenna of the same total physical size but whose characteristics can be reconfigured by manipulation of the relative phase and amplitude weighting applied to each element enables synthetic focusing to an arbitrary point in space via signal processing carried out after the data has been acquired in this piece-wise fashion. This provides a powerful microwave lens which can be focused to an arbitrary location within the breast. This synthetic focusing ability provides the means of imaging small interior features such as malignant tumours. Also, due to the coherent addition of signals obtained from all elements in the synthetic array when focusing to a given point, the signal-to-noise ratio (SNR) of the measurement is improved by a factor N over a single measurement at a single frequency where N is the number of antenna elements in the synthetic array. Furthermore, by making measurements in the frequency domain, one frequency at a time, and then summing up the coherent signals from all antenna elements at all frequencies (to get a time domain response) the signal to noise ratio is further enhanced by a factor F where F is the number of discrete frequencies used.

By coherent addition of signals at the designated synthetic focal point, the imaging device becomes very sensitive to scattered fields located at the focus. The coherent addition is carried out over all antenna locations and at all frequencies. A useful figure of merit is the increase in sensitivity of the imaging device as a result of focussing signals in this way and this is equal to the product of the number of antenna elements with the number of frequencies. This is also equal to the improvement in signal-to-noise ratio over and above a measurement of reflectivity carried out by a single antenna at a single frequency. For the breast imaging device this factor is 161 x 1024 = 164,864 which is equivalent to an improvement of about +52dB. This is more than sufficient to overcome the two-way attenuation of signals in the breast tissue and skin which, at a depth of 5cm at a frequency of 18GHz, is about –40dB. To this end, higher frequencies than 18GHz could be contemplated with a subsequent improvement in resolution in transverse and down-range directions.

In the preferred form frequencies in the range 10GHz to 18GHz are used. In general, attenuation in the breast tissue increases with increasing frequency. The benefit of using higher frequencies is the improved spatial resolution due to the reduced wavelength. The attenuation encountered does not pose difficulties for the preferred form method and apparatus of the invention due to the enhancement in sensitivity (e.g. +52dB) obtained as a result of coherent addition of received signals over a large number of antenna elements (e.g. 1024) along with integration over (e.g. 161) frequencies. Thus, it is believed that the imaging system of the invention can accommodate higher microwave frequencies which enhances the resolution compared to lower-frequency systems.

Also, the nature of electromagnetic scattering from objects small compared to the wavelength, such as the small malignant tumours of interest in breast cancer screening, needs to be considered. Such objects reflect incident energy back to the receiving antenna according to Rayleigh scattering theory. In Rayleigh scattering, the back-scattered power is proportional to the fourth power of the frequency. Therefore, the back-scattered signal from a small embedded object in the breast is 1.8^4 times larger at 18GHz than it is at 10GHz. This is a factor of approximately 10.5 or +10.2 dB. This enhanced scattering at the high-frequency end of the proposed frequency spectrum also helps to offset the increased attenuation in the breast tissue at the higher frequencies.

The preferred form imaging system of the invention is also non-contact and does not require a liquid immersion medium surrounding the breast and antenna system. In addition, the separation between antennas and breast is typically of the order of ten wavelengths at the lowest frequency of operation (about 30cm at 10GHz). This is advantageous over some prior microwave systems which utilise both a liquid coupling medium and have antenna elements either in contact with the breast or in close proximity to it. The motivation for including a liquid medium around the breast is one of impedance matching with respect to the properties of the interior breast tissue. Reflections from the skin layer can be large thereby reducing the amount of energy entering the breast. If the dielectric constant of the liquid medium is similar to that of breast tissue then the amount of microwave energy penetrating the breast is maximised.

The only residual effects that remain are reflections from the skin and attenuation in all media.

It will be appreciated that alternative forms of the imaging system may have antenna element(s) which directly contact the patient's breast or which are coupled to the patient's breasts via a liquid immersion medium, matching layer or matching plate having the appropriate dielectric constant. With a direct contact imaging system, a robot or other scanning mechanism would be arranged to sequentially move the array of antenna elements directly into contact with the breast at each of the predetermined locations. Similarly, with a liquid, layer or plate coupled system a robot or other scanning mechanism would be arranged to sequentially move the array of antenna elements relative to the breast to obtain the radiation information at each of the predetermined locations defining the synthetic aperture relative to the breast. It will be appreciated that there are various coupling configurations possible. For example, the liquid immersion medium could be applied directly to the patient's breasts or alternatively to the antenna elements. Similarly, the matching layer or plate could be fixed relative to the patient's breasts or alternatively fixed relative to the antenna elements.

Although the preferred form focussed synthetic aperture radar imaging system provides excellent sensitivity to the presence of internally embedded objects such as small tumours, further enhancement to the system's sensitivity is possible through the use of additional signal processing to reduce the contribution from skin reflections. The procedure used is to treat the breast interior as a homogeneous lossy medium for which any reflections from the chest wall can be taken as much lower in magnitude than the reflections from the skin layer. This approximation is justified on the basis that the two-way attenuation of microwave energy in the breast tissue significantly reduces the magnitude of any chest-wall reflections as measured at the antenna. With this approximation in place, the breast interior can be regarded as infinitely deep in the z-direction and any reflections that take place are due entirely to the skin-tissue interface. The reflection coefficient from this idealised breast can then be estimated by applying the Physical Optics technique for electromagnetic scattering. The calculated scattered

field is then subtracted from the measured data prior to applying the imaging algorithm. The effects of the skin are thereby significantly reduced by this procedure leaving only the contribution from the breast interior and any embedded features such as tumours.

While the preferred form imaging system has been described as operating in the range of 10GHz-18GHz, the system could be arranged to operate within other higher or lower frequency ranges in the microwave band and the number of discrete frequencies utilised within the selected frequency range can be adjusted to suit design requirements. Furthermore, the imaging system may be arranged to operate at only a single microwave frequency, although this will degrade image quality.

It will be appreciated that the aperture size within which radiation information is obtained can be altered as desired. Further, the number of predetermined measuring locations within the aperture and their respective spacings may be adjusted for specific requirements. For example, the number of predetermined measuring locations within the aperture may be increased to provide more radiation information in order to enhance the quality of the 3D radar image generated.

While the preferred form imaging system has been described in the context of breast imaging, it will be appreciated that other body parts may also be imaged with the system, for example limbs to detect bone fractures or internal bleeding. Further, the imaging system of the invention could be adapted for brain scanning.

It will be appreciated that the imaging system could be provided in the form of a hand-held portable scanning device which could be used in the field by ambulance drivers and the like.

The foregoing description of the invention includes preferred forms thereof. Modifications may be made thereto without departing from the scope of the invention.

CLAIMS

1. A method for generating a three-dimensional image of a body part including the steps of:

scanning to obtain surface profile information for the body part;

transmitting non-ionizing radiation having a frequency or frequencies above approximately 10GHz into the body part and then receiving non-ionizing radiation reflected back from the body part at a multiplicity of predetermined locations defining a synthetic aperture relative to the body part, the radiation being transmitted and received at each of the predetermined locations by moving one or more antenna elements within the synthetic aperture and sequentially operating the or each antenna element;

obtaining radiation information at each of the predetermined locations from the reflected radiation received; and

processing the surface profile information and radiation information to generate a three-dimensional image of the body part, the three-dimensional image having a plurality of image points.

2. An imaging system for generating a three-dimensional image of a body part including:

a three-dimensional profiler arranged to scan the body part and obtain surface profile information;

a radar device arranged to transmit non-ionizing radiation having a frequency or frequencies above approximately 10GHz into the body part and then receive non-ionizing radiation reflected back from the body part at a multiplicity of predetermined locations defining a synthetic aperture relative to the body part to thereby obtain radiation information at each of the predetermined locations, the radar device having one or more antenna elements which are moveable within the synthetic aperture and operable to transmit and receive the non-ionizing radiation; and

a control system arranged to operate the three-dimensional profiler and radar device, and which processes the surface profile information and radiation information to generate a three-dimensional image of the body part, the three-dimensional image having a plurality of image points.

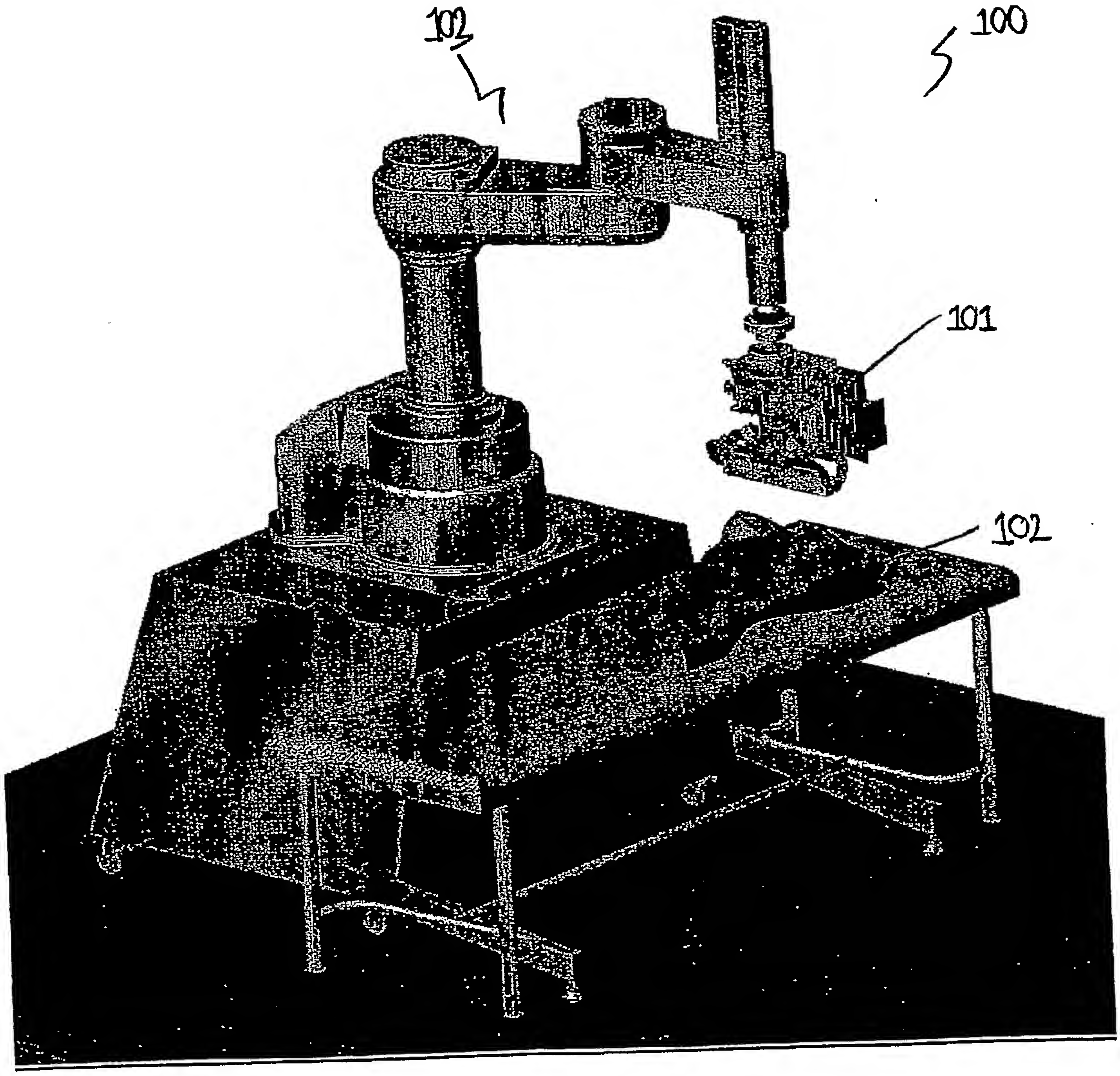


FIGURE 1

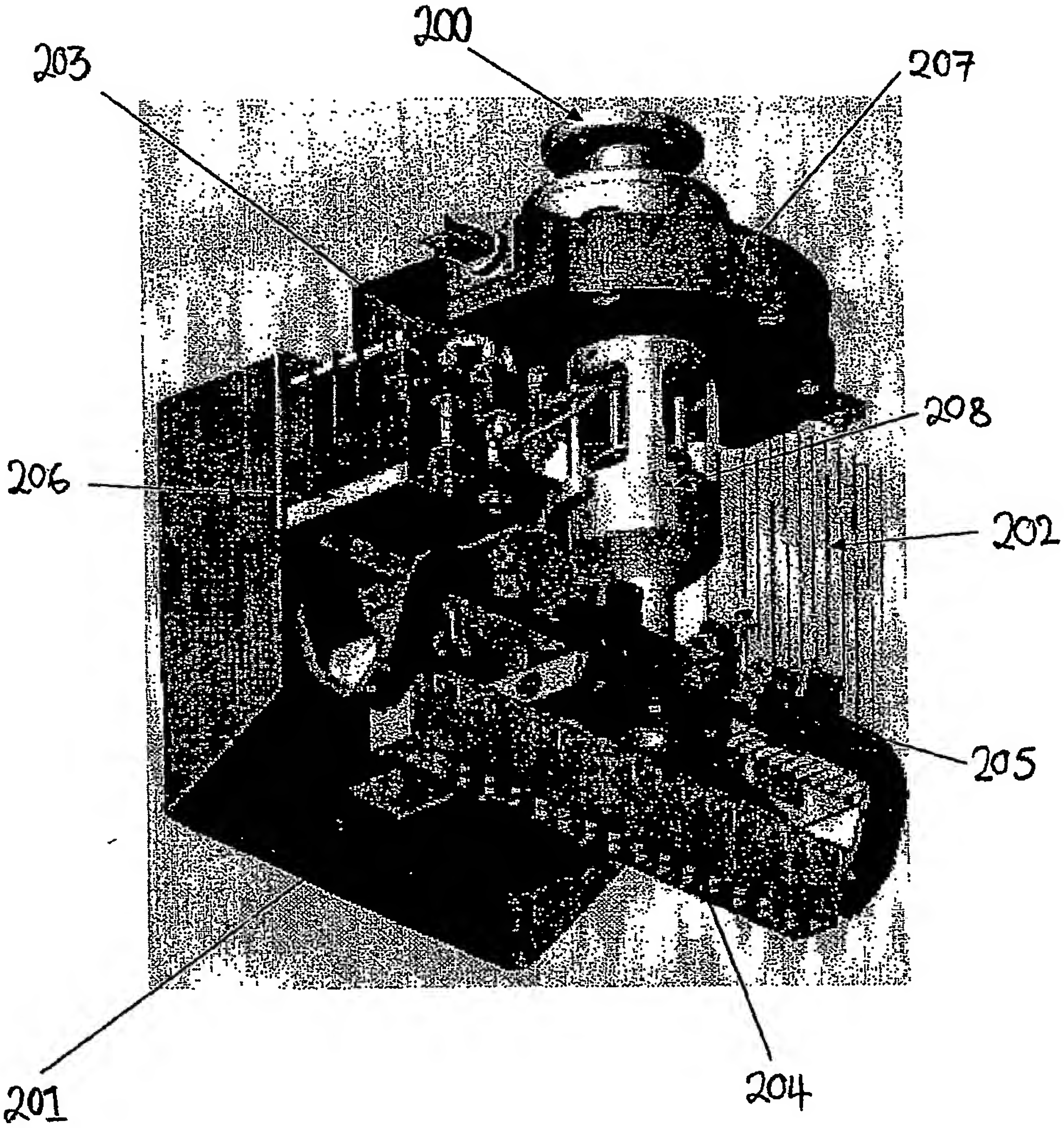


FIGURE 2

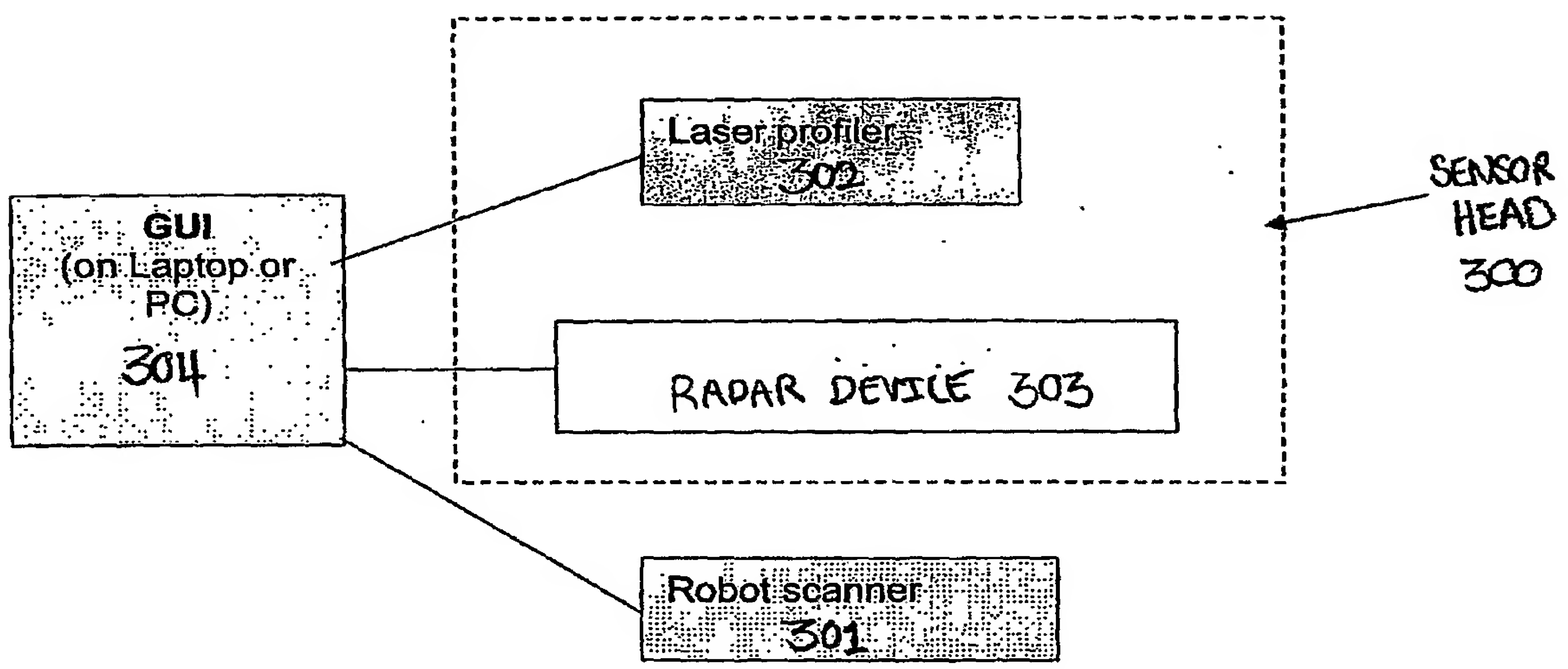

Laser profiler
302
GUI
(on Laptop or
PC)
304
RADAR DEVICE 303
SENSOR
HEAD
300
Robot scanner
301

FIGURE 3

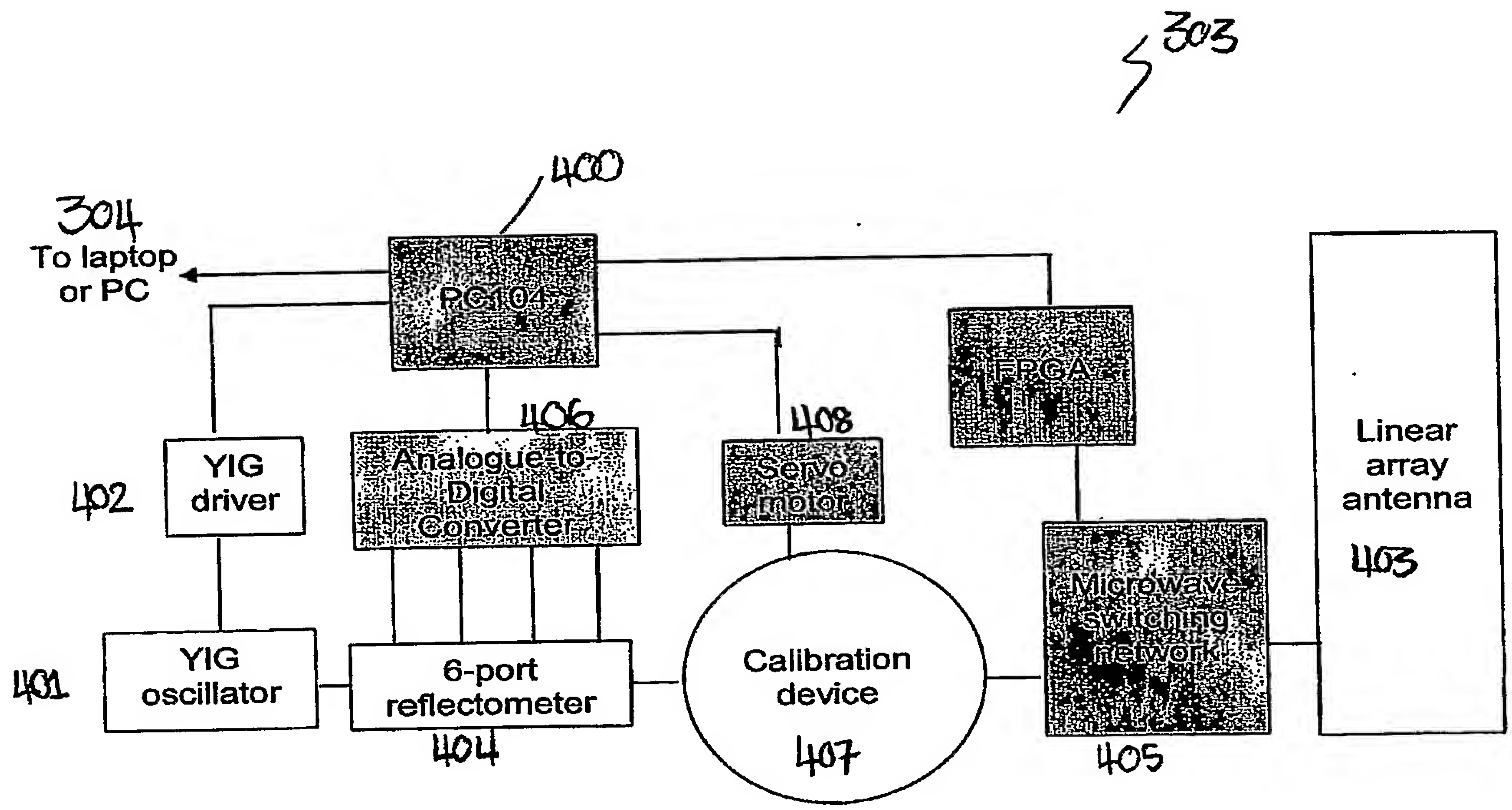
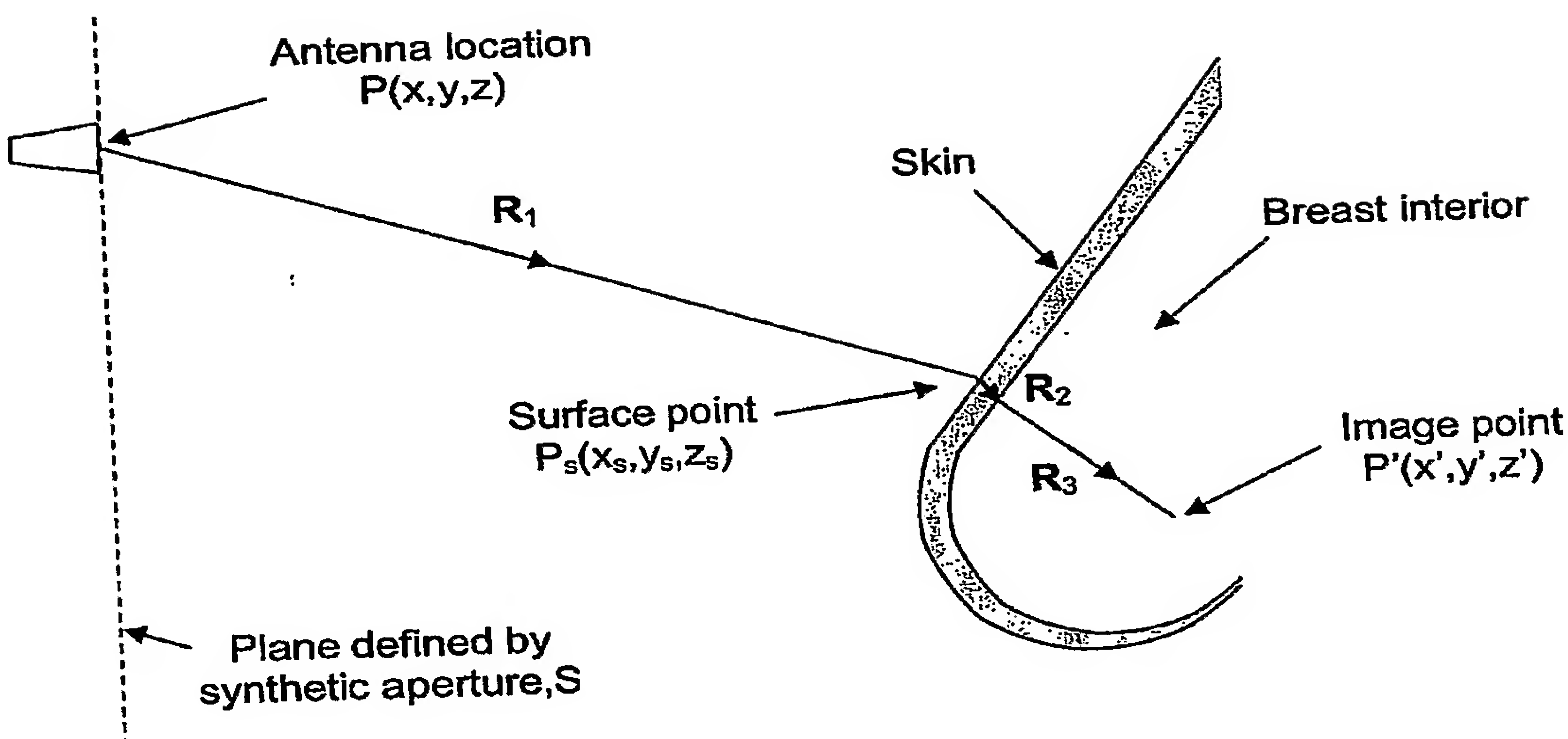

303
400
304
To laptop or PC
PC104
FPGA
Linear array antenna
403
406
408
402
YIG driver
Analogue-to-Digital Converter
Servo motor
Microwave switching network
401
YIG oscillator
6-port reflectometer
Calibration device
407
404
405

FIGURE 4



Antenna location
P(x,y,z)
Skin
R_1
Breast interior
Surface point
$P_s(x_s,y_s,z_s)$
R_2
R_3
Image point
P'(x',y',z')
Plane defined by
synthetic aperture,S

FIGURE 5